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(54) Nuclear camera

(57) A nuclear medicine imaging device includes a radiation camera (10) including a plurality of photo-multiplier tubes (28). Each photo-multiplier tube (28) is configured with an analog to digital converter (30) which converts a detected scintillation event (50) to sampled digital values. A storage device is preloaded with an estimator function which can be derived from a calibration scintillation events. A processor in communication with both the camera (10) and the storage device, detects an event and combines the digital values which are sampled together to arrive at a total area or energy of the scintillation event. Alternately, if a second pulse is detected before the first scintillation event has ended, the area combining of the first event is stopped and a pulse tail is estimated from the estimator functions stored. This estimated tail is then added to the combined data values taken until the time of pile-up. Additionally, the estimated tail is subtracted from the combined data values for the second scintillation event.

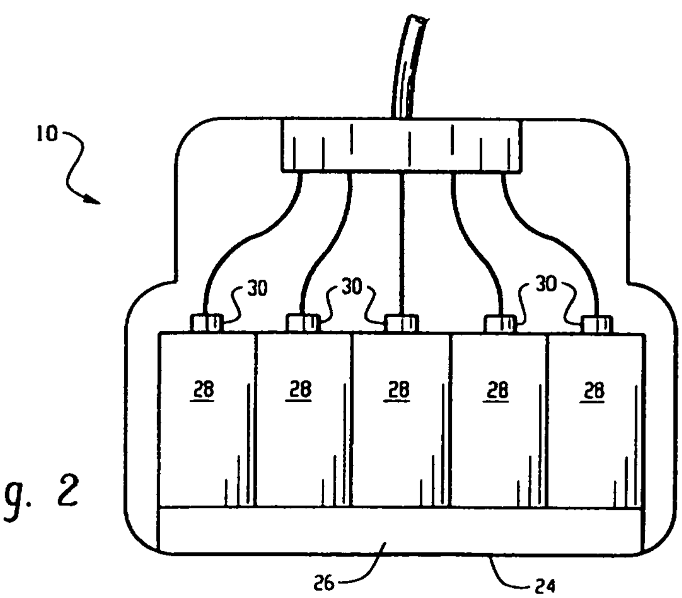

Fig. 2

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Description

[0001] The present invention relates to the field of nuclear medicine, especially for diagnostic imaging. It finds particular application in conjunction with positron emission tomography (PET), and will be described with particular reference thereto. It is to be appreciated that the present invention is amenable to single photon emission computed tomography (SPECT), whole body nuclear scans, positron emission tomography (PET), Compton scattering, other diagnostic modes and/or other like applications. Those skilled in the art will also appreciate applicability of the present invention to other applications where a plurality of pulses tend to overlap, or "pile-up" and obscure each other.

[0002] Diagnostic nuclear imaging is used to study a radio nuclide distribution in a subject. Typically, one or more radiopharmaceutical or radioisotopes are injected into a subject. The radiopharmaceutical are commonly injected into the subject's bloodstream for imaging the circulatory system or for imaging specific organs which absorb the injected radiopharmaceutical. A gamma or scintillation camera detector head is placed adjacent to a surface of the subject to monitor and record emitted radiation. Often, the detector head is rotated or indexed around the subject to monitor the emitted radiation from a plurality of directions. This data is reconstructed into a three-dimensional image representative of the radiopharmaceutical distribution within the subject.

[0003] Each detector head typically includes an array of photo-multiplier tubes facing a large scintillation crystal. Each received radiation event generates a corresponding flash of light that is seen by the closest photomultiplier tubes. Each photo-multiplier tube that sees an event puts out a corresponding analog pulse, pulses from tubes closest to the flash being bigger than pulses from further away tubes. The analog pulses from the individual PMT's are digitized and combined digitally to generate x and y spatial coordinates approximating the location of the scintillation event in the crystal.

[0004] As demands are made for increased patient throughput and improved image quality, the detector heads are subjected to increasing volumes of gamma ray events per second. For example, in a PET scanner in order to obtain about 150 coincident events per second, each detector head typically receives on the order of 2,000,000 events per second. Indeed, one way to increase scanning speed is to increase the number of events per second. Undesirably, as the number of events per second increase, scintillation pulse events begin to overlap to a greater and greater extent leading to pulse loss and other image degradations.

[0005] To accommodate higher count rates, current emission detector heads shorten the pulse tails produced by some of the photo-multiplier tubes, reducing perceived event overlap. This analog technique is known as delay line clipping. This method desirably reduces the effect of pulse pile-up. However it tends to degrade spatial and energy resolution of a gamma camera if a commensurate shortening of integration time occurs. Moreover, scintillation pulses or events that occur without a pile-up are also subject to delay line clipping. In other words, even when the problem of pulse pile-up is not present, the signals are still clipped. This is typically seen as an engineering compromise between count rate and energy and spatial resolution. An additional technique to reduce the effects of pile-up employs the use of an analog signal extrapolation circuits to correct both the initial or first pulse, and a following pulse. Unfortunately, the number of pulses that can be corrected is limited by the number of estimator/amplifier circuits.

[0006] In accordance with one embodiment of the present invention, a nuclear camera includes detector heads mounted for movement around an examination region and a processor for reconstructing signals from the detector heads into an image representation. Each detector head includes a scintillation crystal that converts each received radiation event into a flash of light. The detector heads also include an array of photo-multiplier tubes arranged to receive the light flashes and, in response, generate an analog tube output pulse. A plurality of analog-to-digital converters convert the analog tube output pulse for each photo-multiplier tube to a series of digital tube output values. A processor reconstructs the image representation from the digital tube output values.

[0007] In accordance with another embodiment of the present invention, a method of estimating an energy of an event detected by a medical imaging device includes detecting a first event. The detected event is sampled at a defined sampling rate while samples from the detected event are combined. When a second event is detected, partially coincident with the first event, the combining is ceased. A remainder representing an estimate of combined samples of the first event is estimated following detection of the second event.

[0008] Ways of carrying out the invention will now be described in detail, by way of example, with reference to the accompanying drawings, in which:

FIGURE 1 is a diagrammatic illustration of a nuclear medicine imaging device in accordance with the present invention;

FIGURE 2 is an enlarged view of one of the cameras of Figure 1, cut away partially to reveal internal elements;

FIGURE 3 is a graphical depiction of a digitized pulse from the analog to digital converter;

FIGURE 4 is a graphical depiction of a pulse pile-up event;

FIGURE 5 is an exemplary flowchart which suitably

embodies the present invention;

FIGURE 6 A, B, and C are embodiments according to the present invention of estimating the tail portion of an event in response to detection of a pile-up;

FIGURE 7 is an embodiment of the architecture of the storage device suitable to practice the present invention;

FIGURE 8 is an illustration of an alternate pulse tail remainder estimator function; and

FIGURE 9 is another alternate pulse tail remainder estimator function.

[0009] With reference to FIGURE 1, a nuclear medicine imaging machine includes a number of imaging cameras **10** circumferentially about a region of interest **12**. Typically, an object from which images are desired is injected with one or more radiopharmaceutical or radioisotopes and placed in the region of interest **12**. The presence of these pharmaceuticals within the object produces emission radiation from the object, a certain amount of which is detected by the cameras **10**. The cameras are positionable radially and circumferentially to optimize data acquisition. The heads are angularly indexed or rotated to collect emission data from a plurality of directions. A processor **14** receives the event and head orientation data from the cameras **10** and processes the information into a volumetric image representation defined by radiation received by each camera at each coordinate. The image representation is then stored in an image memory **16** for manipulation by a video processor **18** and display on an imaging display **20**.

[0010] Referring now to Figure 2, each camera has a radiation receiving face **24** facing the area of interest **12**. The receiving face **24** typically includes a scintillation crystal, such as a large doped sodium iodide crystal **26**, that emits a flash or scintillation of light or photons in response to incident radiation. An array of photo-multiplier tubes **28** receive the light and convert it into electrical signals. An analog to digital converter **30** is associated with the output of each photo-multiplier tube **28** for converting its analog, electrical output signal into a series of digital values. The digital outputs of the converters **30** following each scintillation event are processed and corrected to generate an output signal indicative of (1) a position coordinate on the detector head at which each radiation event is received, and (2) an energy value of each event. The energy is used to differentiate between various types of radiation and to eliminate noise and scattered radiation.

[0011] Referring now to Figure 3, a graphical depiction of a digitally sampled scintillation event typically includes a rapidly changing portion **40**, which reaches a peak **42** and a gradual decreasing tail portion **44**. For the processor to determine the energy of the event, the area underneath the curve is determined. The signal is digitally sampled at a rate sufficient to capture an appropriate number of amplitude values. A rate between 40 to 70 MHZ was found to provide a useful number of samples. Artisans appreciate with further reference to Figure 3, that the integration or combination of sample data points is relatively straightforward in the event of a single scintillation event. The combining becomes problematic when several pulses overlap, a condition known as pile-up.

[0012] Referring now to Figure 4, a graphical depiction of pulse pile-up situation is illustrated. As an example, a first event **50** is detected during an imaging procedure. As the digital values $\mathbf{50_1}$, $\mathbf{50_2}$, $\mathbf{50_3}$,---$\mathbf{50_n}$ are forwarded to the processor for combination, the processor is also searching for a second scintillation event. Upon detection of a near concurrent second event **52**, at substantially the same location, the output of the analog to digital converters **30** combine the signal attributable to the second event to the signal attributable to the first event. The processor **14** resolves the two pulses, e.g. by analyzing slope changes, and estimates the missing rail $\mathbf{50_{12}}$-$\mathbf{50_n}$ of the first pulse. To determine the value for the remainder of the pulse tail of the first event, the processor retrieves an estimation function (more fully discussed below), and applies this function to derive an estimate of the complete pulse area of the first event. Those skilled in the art will appreciate that once an estimate is obtained of the remainder of the first event **50**, the remainder can be subtracted from the combined values of the second scintillation event **52** to also yield an effective estimate of that pulse. Accordingly, the ability to estimate accurately the pulse tail upon a pile-up event produces more accurate overall pulse values improving both energy and spatial resolution at any given count rate.

[0013] With reference now to Figure 5, an exemplary flow chart is provided illustrative of the tasks a processor **14** accomplishes. Initially, a clean, single calibration event is sampled **60**. An estimator function is then determined **62** based on the calibration event. The estimator function is stored **64**, for example, in a calibration function storage device **66** (Figure 1). During a subsequent imaging procedure, a first event is detected, and the outputs of the analog-to-digital converters **30** (Figure 2) are sampled **68** at a specified sampling rate. While individual samples are being combined, the system continues to be monitor **70** for a pile-up condition, i.e. a second scintillation event, occurring before the end of the tail of the preceding event. Additionally, the system continues to monitor status of the combination for the first event, as indicated by decision block **72**. In other words, as long as only the first event is detected, its values are sampled and integrated until the end of the event or a pile-up event occurs. Comparison with the calibration event, slope, time amplitude, and the like can be used to determine the end of a pulse.

[0014] In the event no pile-up event occurs, the first event is sampled until it is completed. In the case of only a single event occurring, the area or energy of the scintillation event is determined merely by adding or integrating each of the individual samples. On the other hand, if a pile-up event is detected by the decision block **70**, the integrating is stopped and the estimator function is applied to estimate the area under the cut-off tail. This estimate is combined **74** with the previously integrated sample values, yielding an estimate of the area or energy of the first event.

[0015] Several alternatives are available to derive the estimator function. Referring now to Figures 6A, B, and C, a calibration event is detected and sampled over some period determined to be representative of an entire scintillation event, for example a thirty clock cycle event. For each sample time, ratios are calculated of the area measured at that sample time to the total area of the entire event. Those skilled in the art will appreciate that for the calibration events, sub-total areas are stored for each sample time. Later, at the end of the calibration period, ratios of area at each time to the total area are determined. For example with reference to Figure 6A, the ratio at time T_{13} represents a small portion of the total area under the curve. Differently, the ratio depicted in Figure 6B represents a larger portion of the total area.

[0016] Once ratios have been determined for each sample, these ratios are stored in calibration storage **66** (Figure 1), for example in a format depicted by Figure 6C. During operation, when a pile-up of events is detected, the ratio/estimator function corresponding to the beginning of the second event is retrieved from storage **66**. The inverse of the retrieved ratio is multiplied with the area calculated at the time of the pile-up (i.e. A_1) resulting in an estimate of the remaining area for the pulse (e.g. A_2). At this point an estimate of the entire pulse is determined by adding A_1 and A_2. In other words, the processor performs one multiplication and one addition. Those skilled in the art will appreciate that these ratios may be refined by adjusting them for different operating conditions, such as crystal temperature, or signal processing channel, since variations may exist depending on such properties of the system.

[0017] With reference now to Figure 7, the calibration storage **66** (Figure 1), alternately is configured to include estimated tail areas (i.e. A_2) based on inputs of combined sample area (A_1), time, and other variables. This arrangement collects more calibration data for storage with the benefit of reducing the number of mathematical operations required of the processor, hence increasing speed. In other words, the processor will query the calibration storage **66** and retrieves an area estimate A_2 instead of a ratio. Thus, the processor **26** need only combine the returned estimated value A_2 with the combined sample value A_1 at pile-up time to estimate the scintillation pulse.

[0018] With reference now to the embodiment illustrated in Figure 8, the processor **26** (Figure 1) calculates an estimate of the area of the pulse tail A_2 by assuming the area to be a defined geometric shape, for instance, triangular, as illustrated. Those skilled in the art will recognize that this embodiment may severely underestimate the total pulse area or energy if pile-up occurs early on as indicated by triangle **90**. This error is acceptable for those situations where the pulse rate is sufficiently high to make the storage device **66** (Figure 1) to retrieve estimator ratios or functions infeasible. Alternatively, this method may be refined by reducing the estimated pulse tail to two or more geometric shapes as indicated by Figure 9 areas **94** and **96**.

[0019] One advantage of the described embodiments resides in improved energy and spatial resolution.

[0020] Another advantage is that pulse tail estimation is not limited to fixed integration times. The pulse can be integrated to the maximum integration time when pulse pile-up does not occur.

[0021] Another advantage resides in the elimination of extra extrapolation/amplifier circuits. The correction for pulse pile-up can be completed at the sampling rate of the digital sampling circuitry without extra circuits to handle multiple pile-up events.

[0022] Yet another advantage resides in a calibration procedure using a statistically significant number of pulses to determine the estimator function. The estimator only assumes that the pulse shape is consistent for the correction to be accurate. In other words, no knowledge of the pulse shape is required. Different estimator coefficients can be selected for a given channel's optics or electronics or operating environment, such as temperature.

[0023] The invention has been described with reference to the preferred embodiments. Modifications and alterations will occur to others upon a reading and understanding of the preceding detailed description. It is intended that the invention be construed as including all such modifications and alterations insofar as they come within the scope of the appended claims or the equivalents thereof.

Claims

1. A nuclear camera including a plurality of detector heads (10) mounted for movement around a subject in an examination region (12) and a processor (14) for reconstructing signals from the detector heads into an image representation, each detector head comprising: a scintillation crystal (26) for converting each received radiation event into a flash of light; an array of photo-multiplier tubes (28) arranged to receive the light flashes from the scintillation crystal (26), each photo-multiplier tube (28) being arranged to generate an analog tube output pulse in response to each received light flash; a plurality of analog-to-digital convertors (30), at least one of the analog-to-digital converters being connected with each

photo-multiplier tube (28) for converting its analog tube output pulse to a series of digital individual tube output values, the processor (14) being arranged to reconstruct the image representation from the individual tube output values; and a storage device (66) for storing an estimator function derived from a calibration radiation event, wherein the processor (14) is arranged to integrate the series of digital values corresponding to a first pulse (50) until the first to occur of the end of the pulse and the detection of a beginning of an overlapping second pulse (52) and, in response to detecting the beginning of the overlapping second pulse, the processor is arranged to access the estimator function in the storage device (66) to estimate a remainder (A_2) of the first pulse.

2. A nuclear camera as claimed in claim 1, wherein the estimator function includes a plurality of ratios of portions of the calibration event at selected sampling times the processor comprising: a multiplier for multiplying a ratio corresponding a sampling characteristic of the first event at the time which the second event is detected and the integrated digital values between detection of the first and the second events to produce an estimated remainder; and an adder for adding the integrated digital values and the estimated remainder.

3. A method of estimating an energy of an event detected by a medical imaging device including: detecting a first event (50); sampling the detected event at a sampling rate; combining the samples from the first detected event; detecting a second event (52) partially coincident with the first event (50); ceasing combining the samples in response to detecting the second event; estimating a remainder (A_2) representing an estimate of combined samples of the first event (50) following detection of the second event (52); and for the first event (50), totaling the combined samples of the detected first event (50) and the estimated remainder (A_2).

4. A method of estimating as claimed in claim 3, further including on detection of the second event: combining sampled values to generate a second event combined value; and removing the estimated remainder (A_2) of the first event from the second event combined values.

5. A method of estimating as claimed in claim 3 or claim 4, further including: determining a family of estimator functions for the medical imaging device from calibration events.

6. A method of estimating as claimed in claim 5, wherein the estimator function determining step includes: calculating ratios of combined samples up to selected sampling times within a portion of a calibration event, to combined samples over the entire calibration event.

7. A method of estimating as claimed in claim 5 or claim 6, wherein the estimator function determining step further includes: combining combined calibration samples up to selected times with the corresponding ratios to produce a plurality of remaining area estimates.

8. A method of estimating as claimed in any one of claims 3 to 7, wherein the remainder estimating step includes: based on a sampling time substantially coincident with the detection of the second event, retrieving a corresponding ratio; and multiplying the combined samples and the retrieved ratio.

9. A method of estimating as claimed in any one of claims 3 to 8, wherein the remainder estimating step includes: based on combined samples of the first event (50) before detection of the second event (52), and the detection time of the second event, retrieving the remaining area estimate (A_2).

10. A method of estimating as claimed in any one of claims 3 to 9, wherein the remainder estimating step includes: determining an area under at least two points (94, 96) of an estimated pulse tail curve generated from the calibration events.

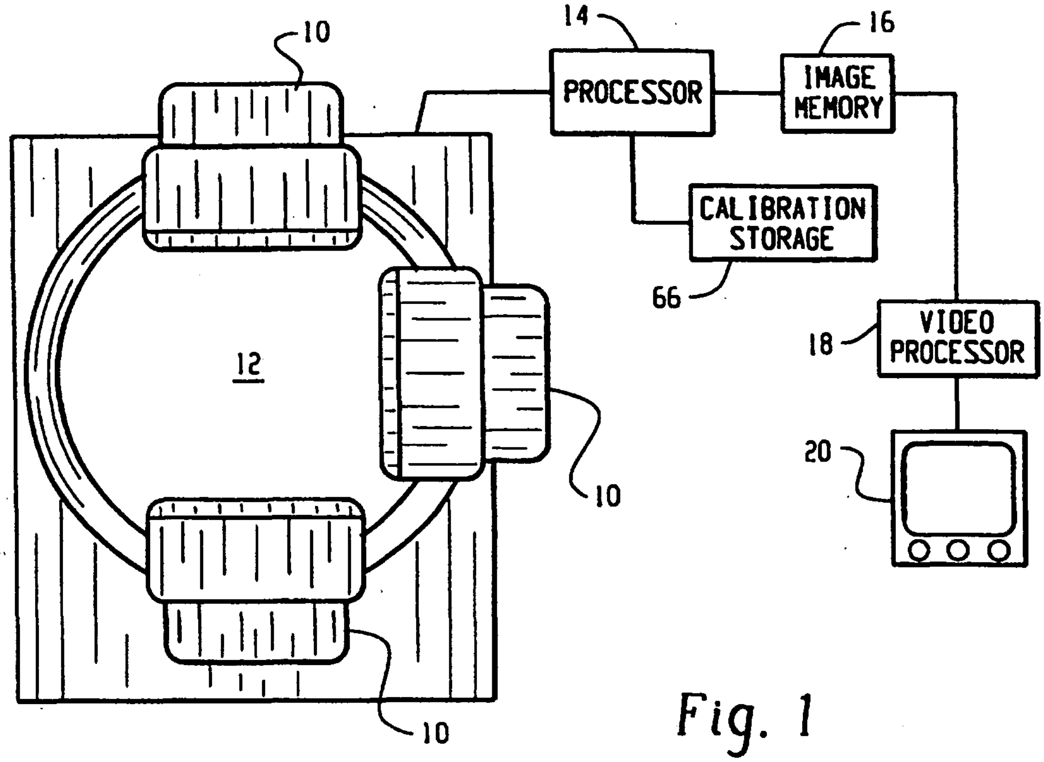
10
14
PROCESSOR
16
IMAGE MEMORY
CALIBRATION STORAGE
66
12
18
VIDEO PROCESSOR
20
10
10

Fig. 1

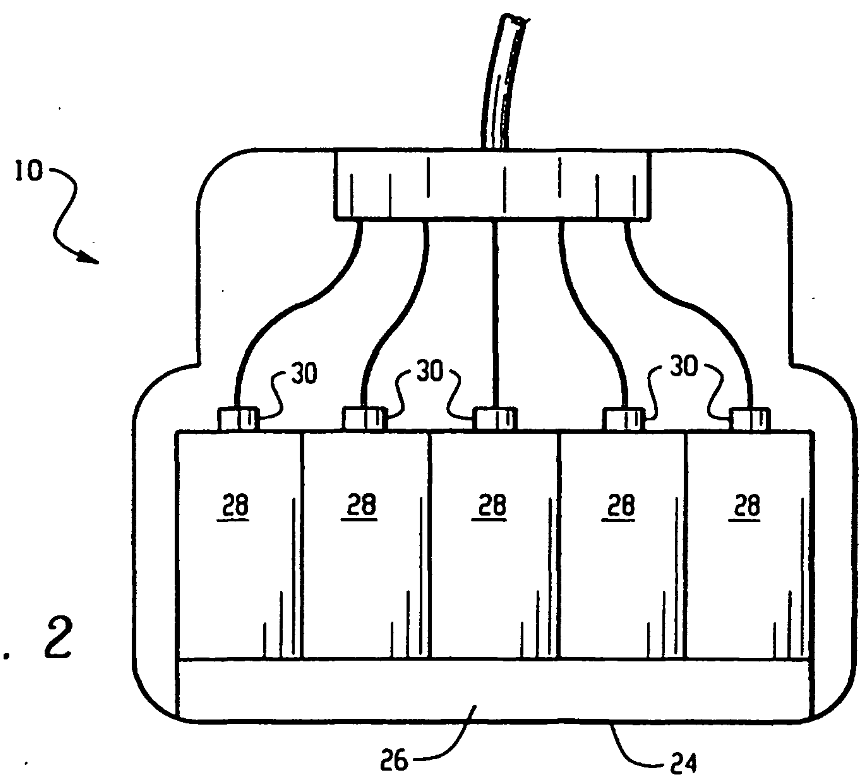
10
30
30
30
28
28
28
28
28
26
24

Fig. 2

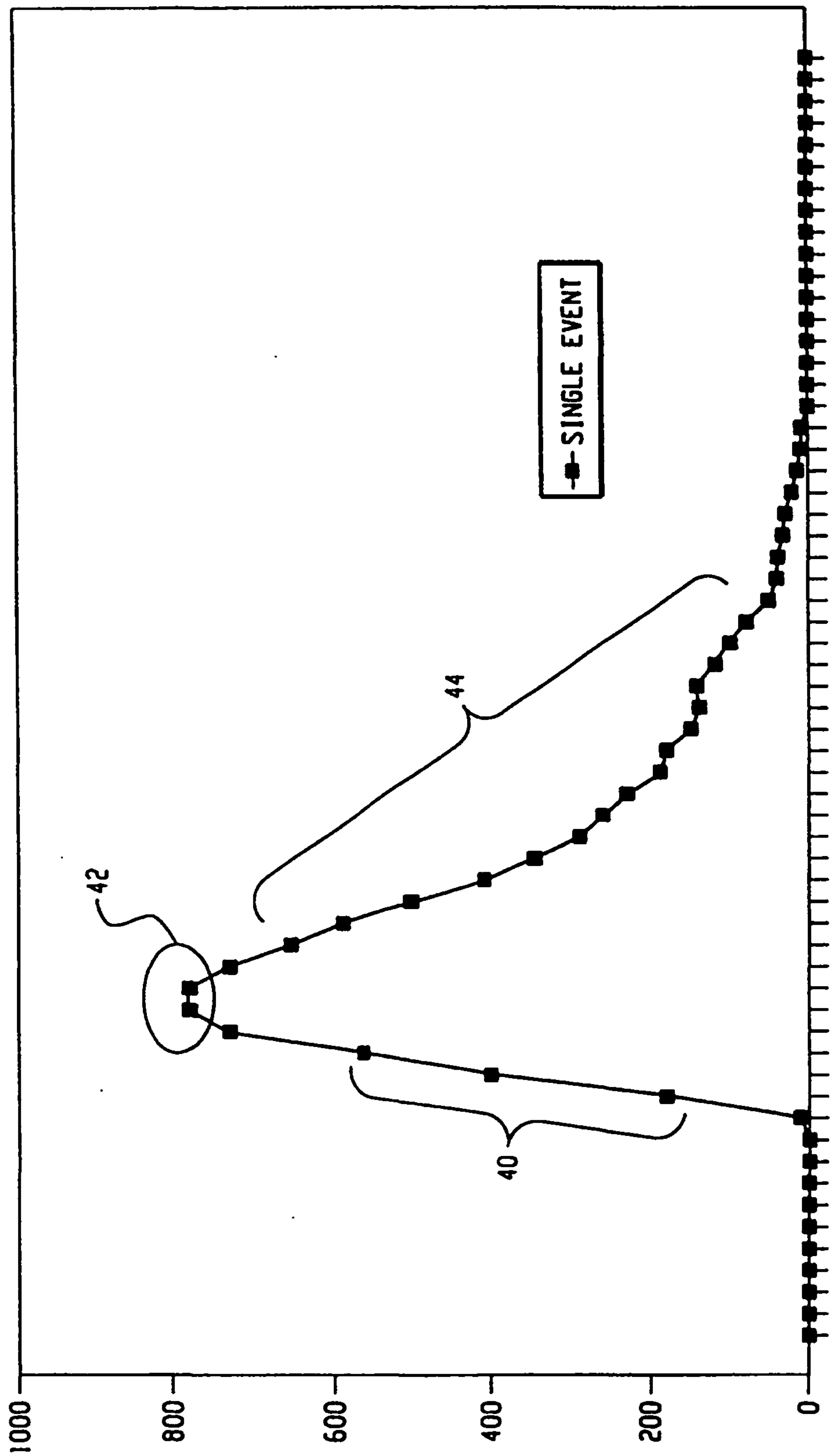

Fig. 3

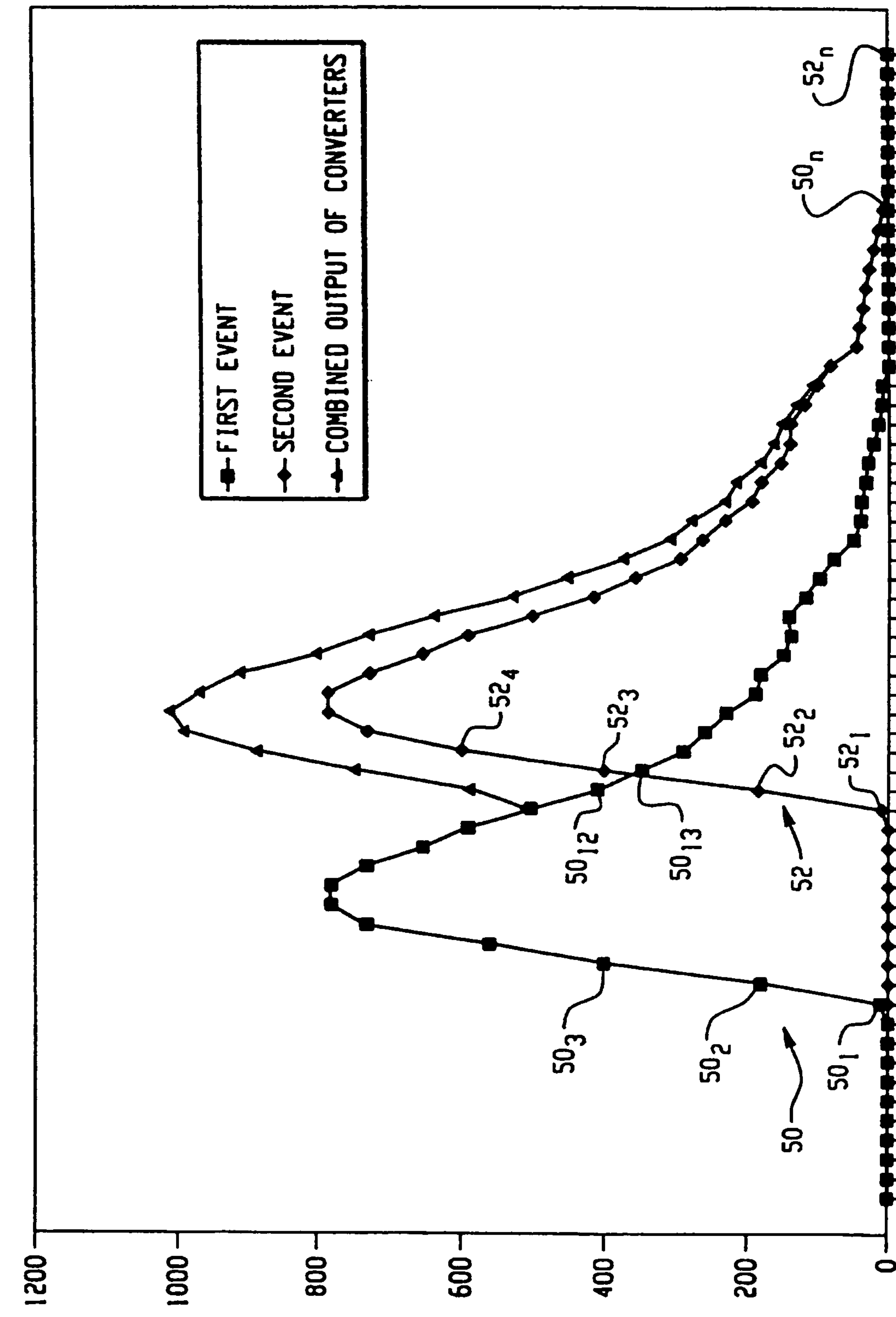
FIRST EVENT
SECOND EVENT
COMBINED OUTPUT OF CONVERTERS
1200
1000
800
600
400
200
0
50_1
50_2
50_3
50
50_{12}
50_{13}
50_n
52_1
52_2
52_3
52_4
52
52_n

Fig. 4

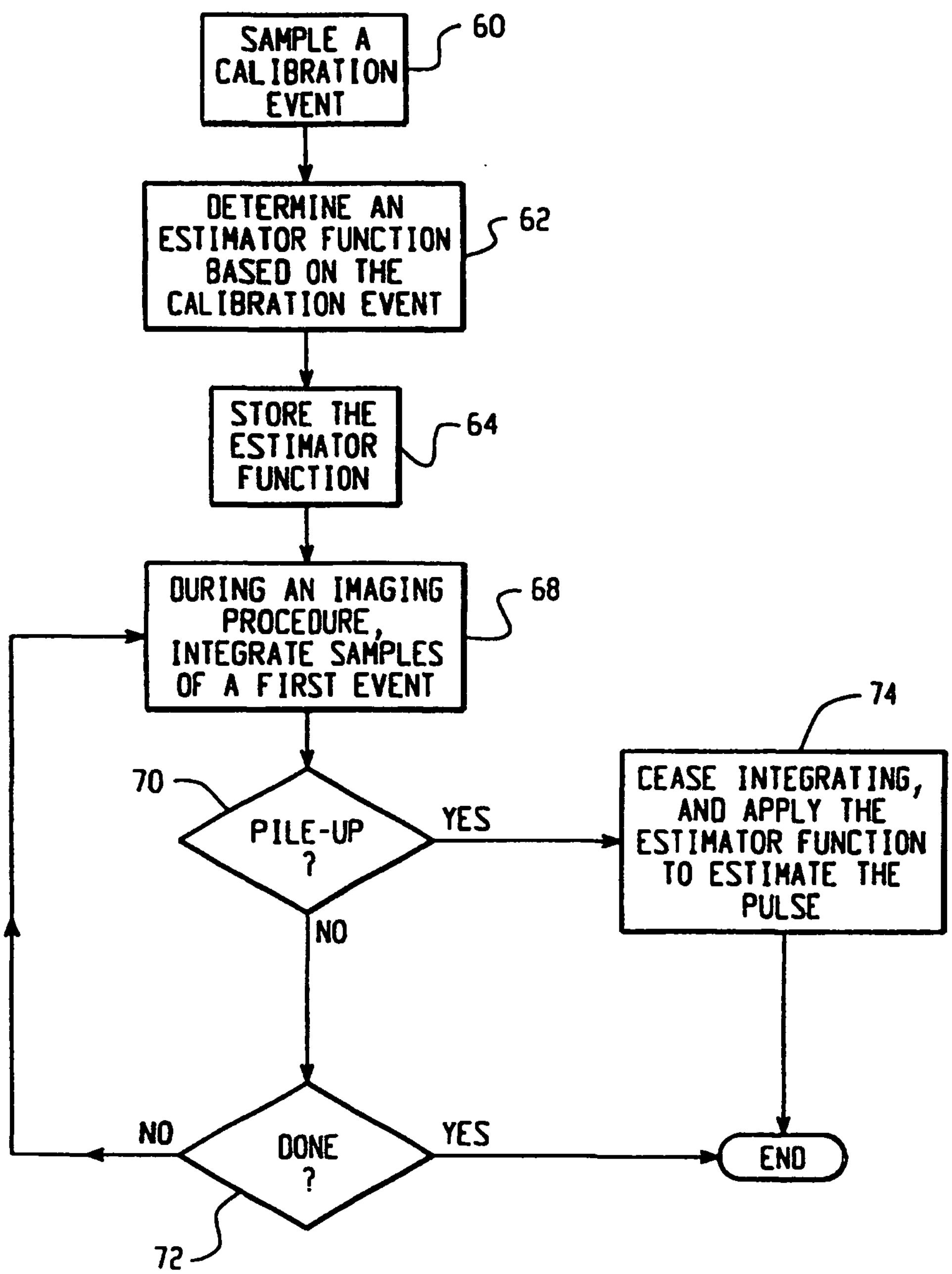
SAMPLE A CALIBRATION EVENT
60
DETERMINE AN ESTIMATOR FUNCTION BASED ON THE CALIBRATION EVENT
62
STORE THE ESTIMATOR FUNCTION
64
DURING AN IMAGING PROCEDURE, INTEGRATE SAMPLES OF A FIRST EVENT
68
70
PILE-UP ?
YES
NO
74
CEASE INTEGRATING, AND APPLY THE ESTIMATOR FUNCTION TO ESTIMATE THE PULSE
NO
DONE ?
YES
END
72

Fig. 5

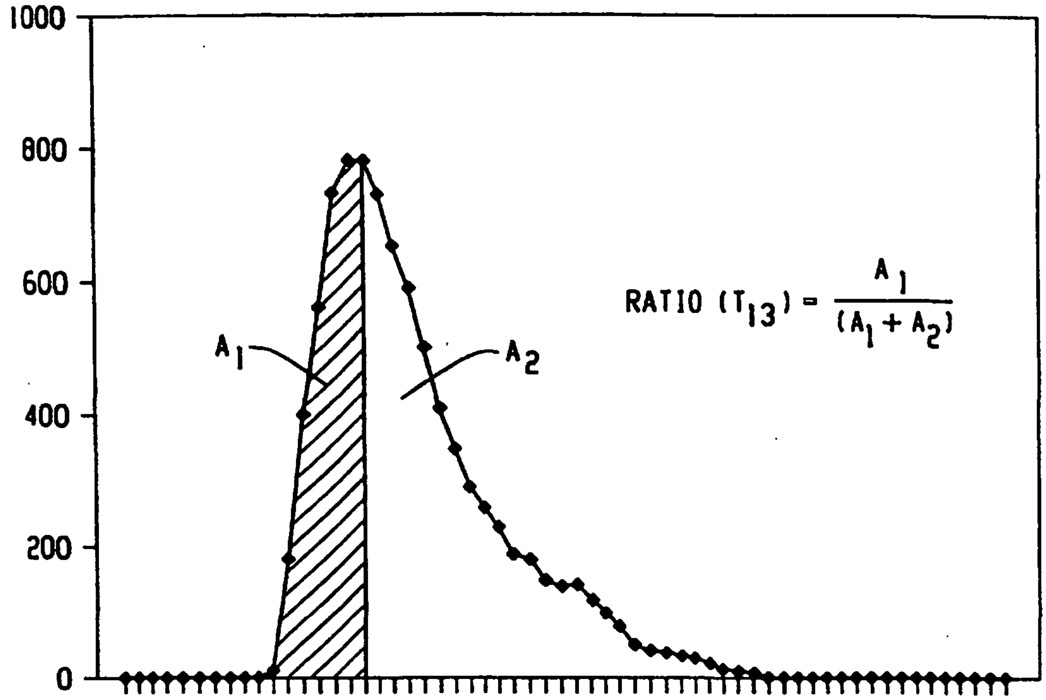
1000
800
600
400
200
0
RATIO $(T_{13}) = \frac{A_1}{(A_1 + A_2)}$
A_1
A_2

Fig. 6A

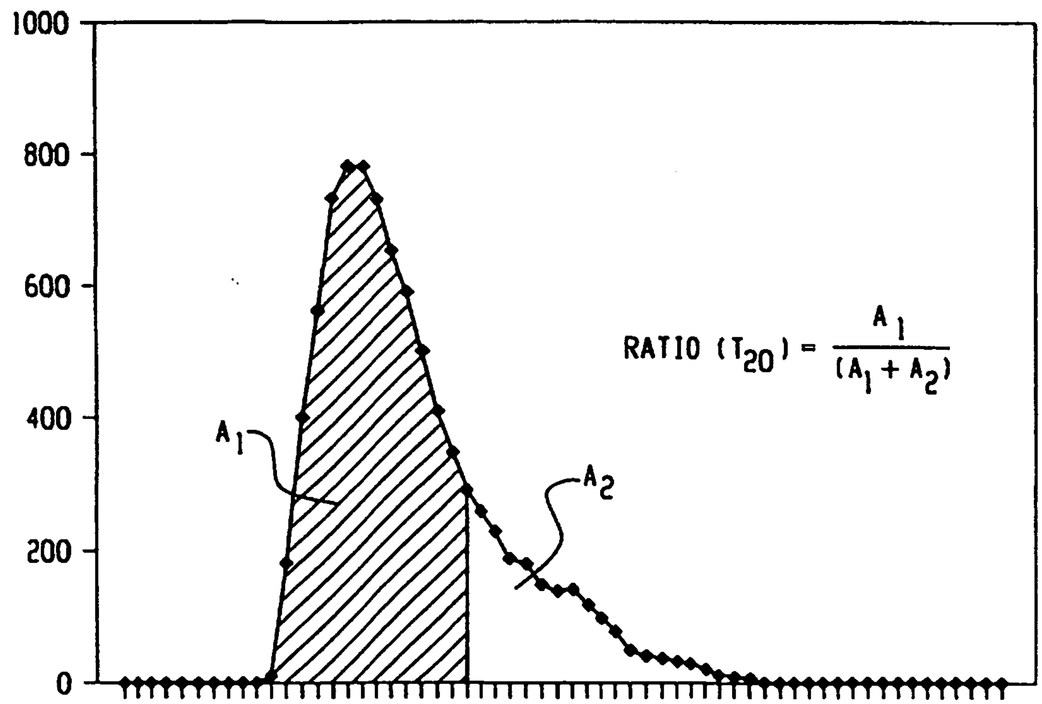
1000
800
600
400
200
0
RATIO $(T_{20}) = \frac{A_1}{(A_1 + A_2)}$
A_1
A_2

Fig. 6B

TIME	1	2	3	4	5	6	7	8	9	10	11	. . .	T-2	T-1	T
RATIO	0.01	0.1	0.2	0.32	0.39	0.41	0.5	0.55	0.6	0.65	0.67	. . .	0.98	0.99	1

Fig. 6C

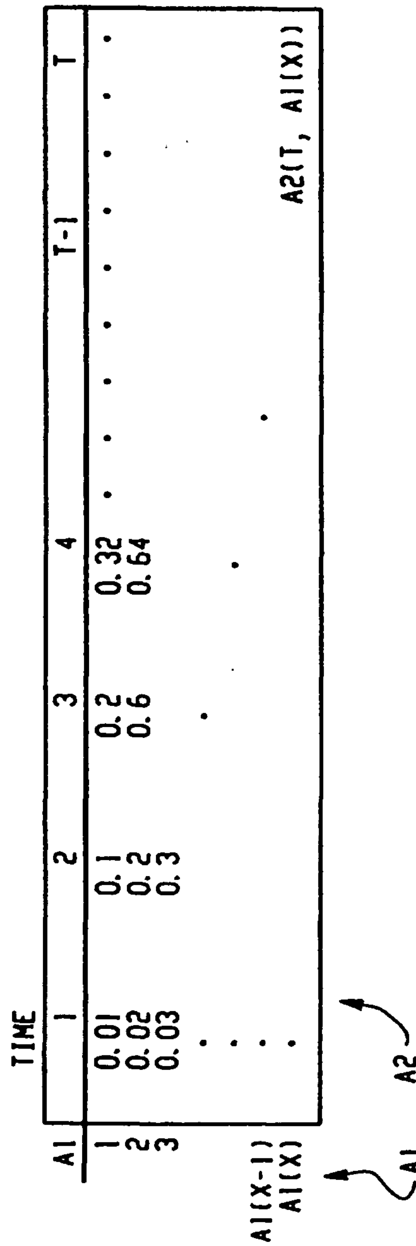

Fig. 7

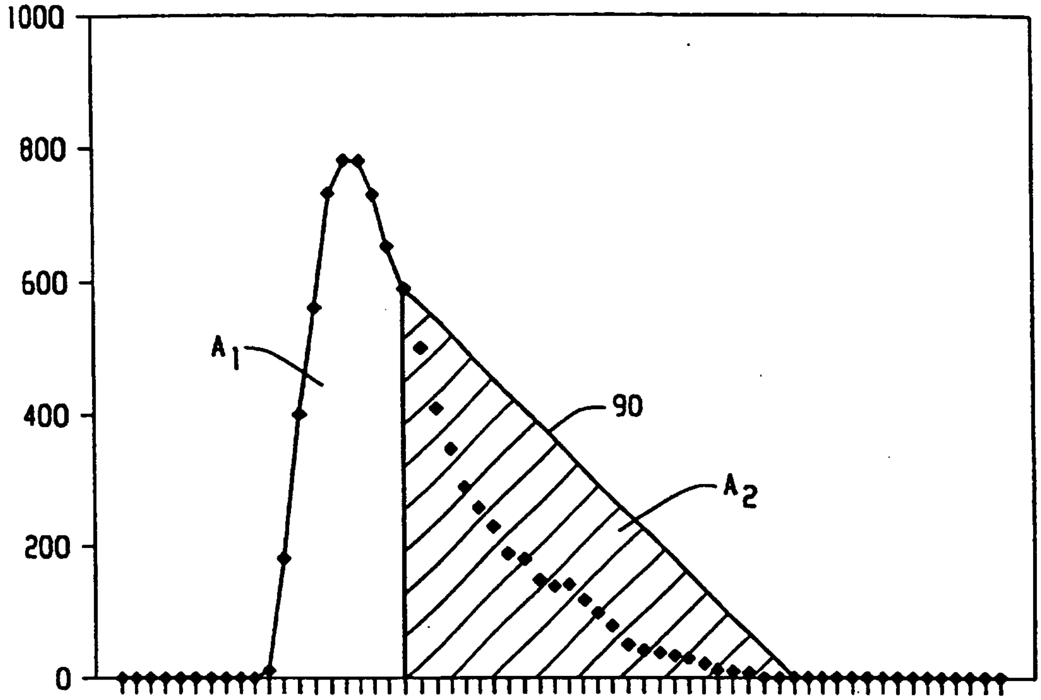

Fig. 8

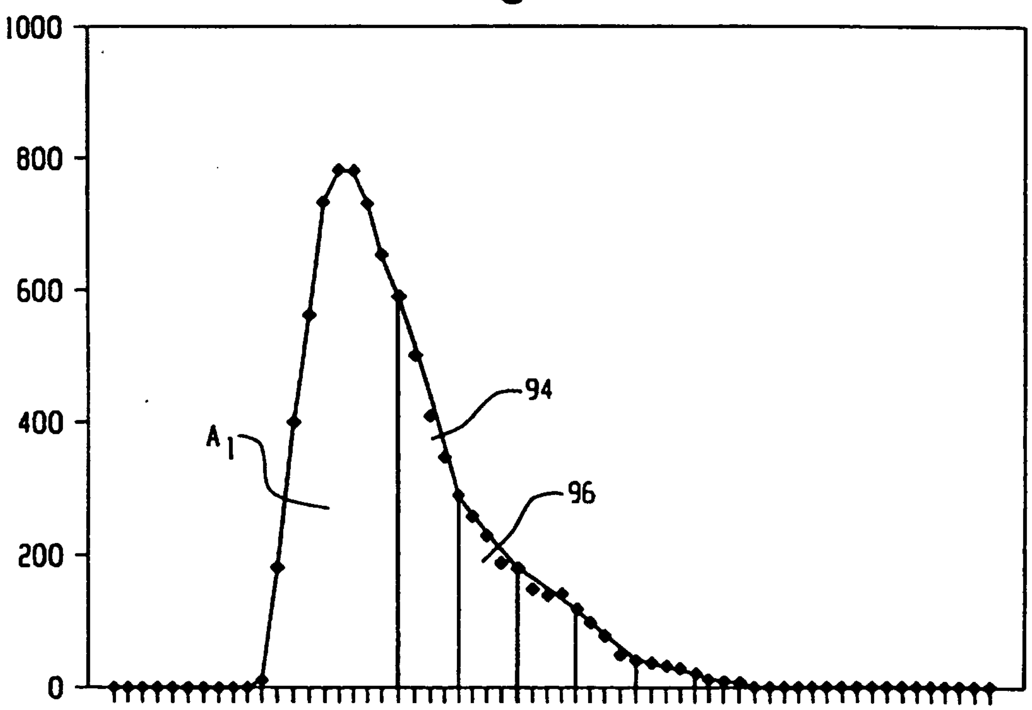

Fig. 9